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# **Adverse condition testing with hip simulators**

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## **Abstract**

Recent clinical failures and abnormal sounds observed in certain prosthetic hip designs have directed attention to adverse condition hip simulator testing. In the present study, hip simulator wear and friction tests were made with a macroscopic separation of the bearing surfaces in the swing phase, steep acetabular cup position, increased load, poor or lacking lubrication, roughened bearing surfaces including titanium third bodies, and with combinations of these. The only conditions that resulted in squeaking were dry sliding with alumina-on-alumina and metal-on-metal, and an extreme peak load of 4 kN with serum lubricated metal-on-metal. The wear rate of metal-on-metal increased by an order of magnitude when the cup position was steep, 64 degrees, compared with the optimal position of 48 degrees. The increase of the peak load from 2 kN to 3 kN with the cup position of 48 degrees increased the running-in wear of metal-on-metal, but the steady state wear rates were equally low, of the order of 1 mg per one million cycles. Crosslinked polyethylene was the superior cup material under adverse conditions, including dry sliding and roughened femoral head.

Keywords: alumina-on-alumina; metal-on-metal; squeak; acetabular cup position

## Introduction

Due to recent clinical failures of certain large-diameter CoCr-on-CoCr hip prostheses [1], more attention has been paid on adverse testing conditions in hip simulators to more closely reproduce the clinical reality where the conditions often are far from optimal. These include steep inclination angle of the acetabular cup [2], laxity of the joint [3], increased loading due to obesity or high-demand activity [4], poor lubrication or dry sliding [5], and roughening of the bearing surfaces [6] for a number of reasons, including titanium third bodies [7]. The principal difficulty is in the anticipation of the clinical relevance of each exceptional condition, or combinations of them. For instance, the percentage of patients with insufficient joint fluid is not known. State-of-the-art bearing couples show minimal wear under normal test conditions, so little that it may be difficult to measure. Under adverse conditions however, early failure may occur. Damage is easy to cause in the laboratory, but it may represent a rare event clinically, in which case it is uninteresting. The steep cup angle is so common in orthopaedics [8] that afterwards it seems odd that steep angle tests were not made with large diameter metal-on-metal (MoM) bearings before their wide introduction in the early 2000s. Steep cup angle is presently known to be tribologically detrimental for these designs [2,9].

Laxity can lead to the separation of the bearing surfaces in gait and in other daily activities. The separation between the femoral head and the acetabular cup has been measured to be of the order of millimetres in gait studies utilizing fluoroscopy [3]. After the heel strike, an impact between the head and the cup takes place in the lax joint. This may be disadvantageous especially with hard-on-hard articulations, as the impact can cause local damage of the contacting surfaces. A click and a squeak are sounds produced by some hard-on-hard articulations [10–12]. The click is likely to result from the impact of the separated bearing surfaces after the heel strike. The squeaking is indicative of ineffective lubrication and high friction which may lead to a wear damage of the bearing surfaces. These sounds are

disturbing but presently it is not known how ominous they really are. The squeak is readily reproduced in dry sliding [13]. A separation mechanism for the HUT-4 hip joint simulator [14] was implemented, and used with alumina-on-alumina (AoA) and MoM. With an ultra-high molecular weight polyethylene (UHMWPE) acetabular cup, the laxity of the joint may not be a risk in the way it is with hard-on-hard articulations, but the increased propensity to dislocation of the lax joint is naturally present irrespective of the bearing materials. A dislocation can damage the bearing surface of the head which has tribological consequences. If the head is not replaced, increased UHMWPE wear by abrasion may follow.

Moreover, MoM tests were run in the present study without separation in the HUT-4 simulator with (a) normal conditions, (b) steep cup position of  $64^\circ$ , and (c) increased peak loads of 3 kN and 4 kN. Friction tests were run with the BRM simulator [13] with several different bearing couples and tribological conditions.

The variety of possible combinations of adverse test conditions and bearing couples is overwhelming. The present study reports several experiments that are considered relevant and interesting, with a view to directing attention to the possibilities and importance of adverse condition testing. It was hypothesized that adverse testing conditions in hip simulators can be predictive of implant behaviour in clinically relevant, sub-optimal conditions, and therefore the possible negative results are worthy to be taken seriously as the earliest warning signals before new prosthetic hip designs are taken into clinical trials, not to mention large scale clinical use.

## Materials and Methods

A separation mechanism was constructed into the HUT-4 hip joint simulator (Figs. 1 to 3) described in detail elsewhere [14]. In the normal walking simulation, the load is implemented by a pneumatic cylinder, a proportional pressure controller and a double-peak input signal. The signal is adjusted so that the maximum load is 2 kN (Fig. 4). After the toe-off, during the swing phase, the cylinder pressure is allowed to exhaust freely. The true load does not however decrease below 300 N before the next heel strike occurs, and no separation takes place. The load is measured with a force transducer fixed to the piston rod. The loading surface of the connecting piece that is pressed downwards by the transducer is horizontal. Between the vertical loading bar of the acetabular component and the connecting piece there is a universal joint that makes the cup self-centring on the head. The connecting piece has one degree of freedom, vertical translation, as it moves along a linear bearing. This assembly functions as the load guide. For separation studies, a constant pressure was applied to the piston rod side to lift the piston together with the force transducer during the swing phase. The gain of the load control was increased so that the maximum load during the load bearing phase was still 2 kN. The connecting piece together with the rail, universal joint, loading bar and the acetabular component were lifted upwards by two vertical springs so that the connecting piece remained in continual contact with the force transducer. The height of the lift could be easily adjusted (Fig. 2). The flange of the loading bar was attached to the acetabular component by two M5 stainless steel screws. The duration of the rise, of the maximum separation, and of the fall were 100 ms each. The duration of the stance phase was 640 ms. The cycle frequency was 1.06 Hz.

The kinematics of the cup from the toe-off to heel strike was dominated by the inferior and superior edge contacts (Fig. 1). Since the distance from the centre of the universal joint to the centre of the prosthetic hip was 140 mm, two orders of magnitude higher than the

separation, the motion of the cup could be considered to consist of horizontal and vertical translations. During the rise, the inferior edge of the cup first impacted against the femoral head and then the cup followed the arc of the inferior edge contact. The steeper the cup angle, the larger the horizontal motion. During the fall, the motion of the cup in horizontal directions was not controlled and momentarily there was no contact between the head and the cup. After the heel strike, the superior edge first impacted against the femoral head and then the fall continued so that the cup followed the arc of the superior edge contact ‘downhill’ until the start of normal articulation, in other words, self-centring, or relocation, took place. Note that the horizontal motion during the rise is likely to move the impact site on the femoral head closer to the centre of contact. No lateral forces were applied to the cup by springs or other means.

The system tests for the new separation mechanism were run with a 28 mm diameter AoA. Its diametral clearance was 0.05 mm. The cup was fixed with a taper into a titanium acetabular shell around which acrylic bone cement was cast so that desired abduction and anteversion angles were achieved. The bearing surface of the cup was sub-hemispherical ( $165^\circ$ ). The head was fixed to a stainless steel holder using a taper fit. In the system tests (Table 1), the joint was immersed in distilled water. In the first actual wear test with separation and serum lubrication, a 50 mm diameter Metasul (Protasul-21WF) MoM bearing with a diametral clearance of 0.16 mm was used (Fig. 3). It was manufactured in the early 2000s by Centerpulse Orthopedics Ltd (Switzerland). The cup was sub-hemispherical ( $165^\circ$ ).

The following MoM tests were run without separation. In the study of the MoM wear with an optimal ( $48^\circ$ ) vs. a steep ( $64^\circ$ ) position of cup using 2 kN peak load, and with an increased peak load of 3 kN (cup position  $48^\circ$ ), Birmingham hip resurfacing (BHR) prostheses of 46 and 54 mm diameter were used. They were manufactured in the early 2000s by Midland Medical Technologies Ltd (U.K.). Their diametral clearances were 0.25 mm and

0.32 mm, respectively. The cups were sub-hemispherical ( $162^\circ$  and  $165^\circ$ ). For the BHR heads, fixation parts were designed for cementing (Figs. 5 and 6). In the test with the 3 kN peak load, one 50 mm diameter Metasul MoM bearing (cup position  $48^\circ$ ) was included in addition to the two BHRs (Fig. 7). It was similar to that used in the separation test. The test duration was 3.3 million cycles. Finally a test with 4 kN peak load was run with 3 similar 52 mm diameter Metasul MoM bearings (cup position  $48^\circ$ ). Their diametral clearance was 0.16 mm and they were similar to those of an earlier study [15] and to the 50 mm Metasuls with the exception that their diameter was 2 mm larger. The components were fixed to the specimen holders of the simulator using bone cement and taper-fit. For the tests with increased loads, simulating obesity, the loading capacity of the HUT-4 simulator was raised with a pneumatic double-piston arrangement operating with a low (0.6 MPa) pressure. The loading beam was stiffened (Fig. 8). The wear in the MoM tests was evaluated from the Co and Cr concentrations of the used lubricant (HyClone Alpha Calf serum SH30212.03 diluted 1:1 with Milli-Q grade distilled water; no additives) by atomic absorption spectroscopy (AAS) and inductively coupled plasma atomic emission spectroscopy (ICP-AES) as described elsewhere [14–16]. The accuracy of the concentration measurements was  $\pm 2$  per cent. Co and Cr were estimated to constitute 90 per cent of the CoCrMo alloy, the density of which was  $8.3 \text{ mg/mm}^3$ . All wear products were assumed to be in the lubricant. At intervals of 550 000 cycles, the test was stopped, the lubricant samples were taken after thorough mixing and measurement of the volume of the used lubricant, the specimens and their holders were cleaned, and the test was continued with fresh serum. The wear at the first 550 000 cycles was considered the running-in wear. After that, the wear rate was determined by linear regression from the 6 points of wear (mg) vs. number of cycles (million), and the goodness of fit  $R^2$  was calculated.

With crosslinked UHMWPE cups, a variety of adverse condition trials were made with



the HUT-4 so that one or more conditions were exceptional, including steep cup angle, separation, roughened head, distilled water lubrication, Ringer's solution lubrication, and dry sliding. The wear was quantified by weighing the cup. Moreover, the effect of titanium third bodies was tested in the HUT-4 with AoA, as it has been assumed that titanium third bodies may be one cause for squeak clinically [7]. The particles that were put between the bearing surfaces were made using a file from a titanium acetabular shell.

Measurements of frictional torque  $T$  were made with the biaxial rocking motion (BRM) hip joint simulator. The device and the method of measurement have been described elsewhere [13]. With respect to the nomenclature of [13], the frictional torque in the present study was the sum of the vertical and leaning components, that is, the total torque. The load was static 1 kN. The running time was usually 2 h, at the end of which a steady state  $T$  value was recorded. The bearing surfaces of a 28 mm diameter Biolo Forte AoA joint were deliberately roughened with emery paper to an  $R_a$  value of 0.1  $\mu\text{m}$  (Fig. 9). A 50 mm diameter Metasul MoM joint was first run in water, then dry and finally in serum. The crosslinked polyethylene cup (Durasul) was described in an earlier study [17]. It was run against smooth and roughened ( $R_a$  0.1  $\mu\text{m}$ ) CoCr (MoP) and alumina (AoP) heads of 28 mm diameter, lubricated and dry. The surface roughness was measured with a diamond stylus instrument using 0.25 mm cutoff and 1.25 mm evaluation length. At least 10 measurements were made from each component.

## Results

In the running of the separation tests with AoA and MoM, the sound of click dominated. Squeak was absent. The sound depended on the course of the cup from maximum separation to normal articulation. The maximum sound was produced when the position of the cup was not controlled horizontally in the swing phase, which was the normal test procedure. If

however the cup was pushed medially so that the superior edge was in contact during lift, maximum separation and descent, the click was absent. Despite the impacts, the measured load profile was as smooth as without separation. The impact of the superior edge of the alumina cup caused only a minor streak-like mark by grain removal on the alumina femoral head (Fig. 10). The load bearing surfaces were not damaged. The total wear of the MoM bearing with the steep position of the cup and separation was 0.4 mg at 0.57 million cycles and 1.1 mg at 1.1 million cycles. Only mild scratching and a streak-like indentation mark by plastic deformation on the femoral head could be observed (Fig. 11).

The HUT-4 MoM tests without separation of 3.3 million cycle duration were uneventful with the exception of the 4 kN test that was discontinued at 1.3 million cycles because of serum darkening, overheating above 40 °C, quick evaporation and loud squeak (Table 2, Fig. 12). With the BHRs, the steep cup angle caused an increase of wear rate by an order of magnitude (Fig. 13). The increase of the peak load from 2 kN to 3 kN increased the running-in wear substantially, but the steady state wear was not increased.

In the BRM simulator, with dry sliding of smooth alumina-on-alumina the squeak was loud, but no damage occurred (Table 3). With dry sliding of roughened AoA a grinding sound dominated and some squeak was heard as well. In water, only grinding was heard with roughened AoA, and this sound diminished as the lubricant was changed to serum. With MoM in water, the frictional torque  $T$  increased from 0.5 Nm to 5.5 Nm within minutes and scratching and galling occurred (Fig. 14). When the running was continued dry for 10 min,  $T$  dropped to 3.9 Nm. As the running was continued with serum,  $T$  further dropped to 1.6 Nm and it was stable for the duration of the test (2 h) although the serum darkened. Grinding took place in all three test conditions, and in addition squeak was heard in water lubricated and in dry sliding. In the reference tests with a crosslinked UHMWPE cup, no sounds were produced.

In the HUT-4 trials, crosslinked UHMWPE showed considerable wear when more than one condition was exceptional. For example with a roughened CoCr head ( $R_a$  0.1  $\mu\text{m}$ ) of 28 mm diameter and lack of proteins in the lubricant (distilled water), wear was 35 mg after 550 000 cycles, corresponding to a wear factor of  $2.3 \times 10^{-6} \text{ mm}^3/\text{Nm}$ . Wear debris was in the form of macroscopic flakes. Heavy polyethylene transfer to the CoCr head took place further increasing the roughness. When a steep cup angle ( $64^\circ$ ) was added to the above, a dislocation occurred quickly and the test had to be discontinued. When Ti particles were put between AoA, squeak occurred in water, but it ceased when lubricant was changed to serum (Fig. 15).

## Discussion

Separation mechanisms have been added to hip simulators earlier [18,19] but they differ from the present mechanism in one important feature. In the earlier mechanisms the separation takes place by forcing the cup in the medial direction by springs which leads to partial separation guided by the superior edge contact during a low-load swing phase. In this way, there is not merely a continual contact between the superior edge and the head during the swing phase, as suggested by fluoroscopy [3], but the edge is intensively rubbed against the head, because the vertical load is not negligible, and so there is a considerable contact force between the edge and the head. In a hip simulator MoM test (39 mm diameter head, cup inclination  $60^\circ$ ), separation caused by a lateral force resulted in a 17-fold increase in the wear rate compared with standard gait conditions [20]. The present mechanism reproduced the impact damage of the femoral head [12] although there was no horizontal loading. In the running with the present mechanism it was found that if the superior edge separation was prevented by pushing the cup medially, as in [20], the click ceased. This would be against the purpose of the study, that is, to reproduce the noise phenomena observed in vivo. Moreover, although the typical in vivo separation to be simulated is of the order of millimetres [3], the

earlier tests are for some reason called ‘microseparation’ studies [18–20] which appears misleading. If the in vivo separation was of the order of micrometers, it would hardly be noticeable, and unlikely to cause any problems tribologically. Similarly, the term ‘stripe wear’ that has been used in ‘microseparation’ studies [18–20] is unknown to tribology as a wear mechanism. Apparently it rather refers to the shape of the damage observed on the femoral head.

Wear of conventional UHMWPE inserts of total hip prostheses has been found to be insensitive to the inclination angle of the cup, 48° vs. 60° [21]. With large-diameter MoM however, the wear increased steeply with increasing angle [15]. In the HUT-4 tests with separation this was not the case with AoA, probably because the contact area was small, due to the high elastic modulus, and did not extend to the edge of cup even with the 64° inclination angle. The edge impact caused a streak-like mark by grain removal on the head (Fig. 8), similar to that known to take place in vivo [12], but this did not lead to further damage. Surprisingly, the separation was advantageous in serum lubricated MoM with steep cup angle apparently due to generous lubricant ingress during the swing phase. The wear rate was very low, and close to that measured for an optimal position without separation [14]. There was no running-in phenomenon, and the total wear after 1.1 million cycles was 1.1 mg only.

The HUT-4 tests with the BHR prostheses in the normal conditions (48° cup position and 2 kN peak load) showed running-in wear values three times higher than 50 mm diameter Metasul tests in similar conditions [14]. This is likely to be attributable to the higher clearance of the BHRs. The steady state wear rate values were equally low. The steep position of the BHR cup resulted in the increase of wear rate by an order of magnitude compared with the optimal position. Similarly, in another MoM hip simulator study (28 mm head diameter), the increase of the cup inclination angle from 45° to 55° resulted in a fivefold increase of the

wear rate [22]. With a 39 mm diameter MoM, the increase of the cup inclination angle from 45° to 60° resulted in a ninefold increase of the wear rate [20]. With the steep angle the contact is bordered by the edge causing a failure of the lubrication film, and metal-metal contact. The same phenomenon was earlier shown with the Metasul couple [15], and it is in agreement with clinical observations [2,9]. Increasing the peak load from 2 kN to 3 kN increased the running-in wear but the steady-state values were equally low as the contact did not reach the edge of the cup, and the lubricant film obviously functioned well after the bedding-in phase.

According to biomechanical studies the peak load in the hip joint in normal level walking is 2.5 times the body weight [23]. An average patient can be assumed to weigh 800 N. Therefore the peak load value of 2 kN has been systematically used in the HUT-4 hip joint simulator. The peak load of 3 kN would simulate a patient weighing 1200 N, that is, suffering from obesity. It has been estimated that in stumbling, the hip peak force can be as high as 11 kN [4]. The peak load of 4 kN was shown to be excessive for a serum-lubricated 52 mm diameter MoM in continuous testing. Serum overheated, lost its lubrication capability after which squeak occurred, and the wear rate was very high. The three simultaneous 4 kN MoM tests showed large differences in the wear rate (Table 2) which is typical of a situation where the limit of performance is being reached. Similar high variability of MoM wear was observed in another hip simulator study with 2.5 kN peak load [24]. Crosslinked UHMWPE cups however have been shown to remain functional even under excessive loading [25].

As a limitation of the present study it may be noted that the AAS and ICP-AES methods of measuring the Co and Co concentrations [14–16] were insensitive to the fact of whether the metals were in ionic or particulate form in the used serum lubricant because of the acid digestion of the samples prior to the analysis. The total concentration was determined, but particles were not studied.

Interestingly, water lubrication resulted in the lowest and the highest  $T$  values ever measured with the BRM simulator. In an earlier friction study with the BRM simulator [13], the frictional torque with a 28 mm diameter BioloX Forte AoA (1 kN static load) in distilled water was only 0.03 Nm, which is the lowest value ever measured with the device. In serum,  $T$  was 0.17 Nm. The highest  $T$  value was that of water lubricated MoM which was severely roughened in the test within minutes by abrasive and adhesive wear, and  $T$  increased from 0.5 N m to 5.5 N m accompanied by sounds of squeak and grinding. Even in dry sliding  $T$  was lower, 3.9 Nm, despite the fact that the bearing surfaces were damaged. In brief, water lubrication worked extremely well with AoA, but did not work at all with MoM. The presence of proteins is a necessity for MoM. The serum lubricated  $T$  value of the MoM with undamaged bearing surfaces was 0.5 Nm [13]. Dry sliding, audible as a squeak, increased friction considerably with AoA and MoM. This finding is in agreement with another hip simulator study [26]. A roll/slide mechanism between the head and the cup has been proposed for the squeaking of AoA [27]. However, in pin-on-disc studies the difference in friction between water lubricated and protein lubricated ceramic-on-ceramic was much smaller than in the BRM, probably due to the absence of the clearance effect [28]. After the deliberate roughening of the AoA bearing surfaces,  $T$  was as high as 2.7 Nm in dry sliding, 2.1 Nm in distilled water, and 1.6 Nm in serum. A grinding sound dominated in all three conditions and in dry sliding only some squeak was additionally produced. With an UHMWPE cup, no sounds were ever produced, and even dry 2 hour tests against roughened heads could be run without problems. Frictional heating was substantial of course but the bearing remained functional. If there is doubt of in vivo lubrication, the crosslinked UHMWPE cup seems to be the safest choice.

## Conclusions

Adverse condition wear and friction tests were run with hip simulators for metal-on-metal, alumina-on-alumina, metal-on-polyethylene and alumina-on-polyethylene. The tests included macroscopic separation of the bearing surfaces in the swing phase, steep acetabular cup position, increased load, poor or lacking lubrication, and roughened bearing surfaces.

The laxity of the joint together with a steep acetabular cup position did not result in squeak with lubricated AoA or MoM articulations. A streak-like mark on the femoral head was formed by the impact of the superior edge of the cup, which however did not lead to further damage. Reproducing the sound of click that has been reported clinically required a separation of the superior edge from the head.

Dry sliding with AoA and MoM systematically resulted in squeak in the present tests. In serum lubricated tests with deliberately roughened AoA and MoM surfaces, a grinding sound dominated instead of a squeak, and friction was substantially increased. In serum lubricated AoA, titanium third bodies did not lead to squeak.

A steep position of the cup, 64°, with the BHR MoM resulted in an increase of wear rate by an order of magnitude compared with the optimal position of 48°. No difference in the BHR steady state wear rate could be observed between the peak loads of 2 kN and 3 kN, but the running-in wear with the latter value was substantially higher.

A peak load of 4 kN resulted in serum overheating and loud squeak with the 52 mm diameter Metasul MoM. The wear rate was two orders of magnitude higher compared with the 3 kN peak value.

The best material under adverse conditions was crosslinked polyethylene, but even for this material, three adverse conditions simultaneously can be unbearable, e.g., steep cup angle together with roughened head and absence of proteins.

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Table 1. Tests with swing phase separation in HUT-4 hip joint simulator. Peak load 2 kN.

Test no.	Type of test	Type of articulation	Diameter (mm)	Lubricant	Cup inclination (degrees)	Vertical separation (mm)	Test length (10 <sup>6</sup> cycles)
1	system	AoA	28	distilled water	48	0.5	0.2
2	system	AoA	28	distilled water	64	0.6	3.0
3	system	AoA	28	distilled water	64	1.0	1.0
4	wear	MoM	50	serum, diluted 1:1	64	1.0	1.1

Table 2. Wear of MoM prostheses in serum-lubricated HUT-4 hip simulator tests without separation. Test length was 3.3 million cycles with the exception of the 4 kN test which was terminated at 1.3 million cycles.

Prosthesis	Diameter (mm)	Test type	Running-in wear (mg)	Wear rate (mg/10 <sup>6</sup> cycles)	R <sup>2</sup>	Wear factor (mm <sup>3</sup> /Nm)
BHR	46	Normal	14.0	3.6	0.9561	$8.9 \times 10^{-9}$
BHR	54	Normal	12.9	1.4	0.7722	$3.1 \times 10^{-9}$
BHR	46	64°	16.6	33.5	0.9626	$8.4 \times 10^{-8}$
BHR	54	64°	15.1	11.2	0.8164	$2.4 \times 10^{-8}$
BHR	46	3 kN	24.1	1.3	0.9985	$2.2 \times 10^{-9}$
Metasul	50	3 kN	13.6	1.3	0.9476	$2.1 \times 10^{-9}$
BHR	54	3 kN	32.2	2.1	0.6951	$2.9 \times 10^{-9}$
Metasul #1	52	4 kN	18.8	116.0	0.9912	$1.3 \times 10^{-7}$
Metasul #2	52	4 kN	9.3	24.0	0.9988	$2.7 \times 10^{-8}$
Metasul #3	52	4 kN	12.8	86.6	0.9635	$9.6 \times 10^{-8}$

Table 3. Frictional behaviour of different bearing couples evaluated with BRM simulator using static 1 kN load.

Articulation	Diameter (mm)	Condition of surfaces	Lubricant	$T$ (N m)	Sound
MoP	28	original femoral head	serum	1.5	—
			water	1.4	—
			none	2.1	—
MoP	28	roughened femoral head	serum	1.4	—
			water	1.3	—
			none	2.0	—
AoP	28	original femoral head	serum	0.9	—
			water	0.9	—
			none	2.4	—
AoP	28	roughened femoral head	serum	1.1	—
			water	1.4	—
			none	2.2	—
AoA	28	original	serum	0.17 <sup>a</sup>	—
			water	0.03 <sup>a</sup>	—
			none	2.0	squeak
AoA	28	roughened head and cup	serum	1.6	faint grinding
			water	2.1	grinding
			none	2.7	squeak and grinding
MoM	50	original	serum	0.5 <sup>a</sup>	—
MoM	50	damaged in testing <sup>b</sup>	serum	1.6	grinding
			water	5.5	squeak and grinding
			none	3.9	squeak and grinding

<sup>a</sup> Ref. [13]

<sup>b</sup> See Materials and Methods, and Results

## Figure captions

Figure 1. Schematic of separation mechanism designed for HUT-4 hip simulator. Since laxity of joint is simulated, no lateral force is applied. Superior edge separation is necessary for audible click to be produced after heel strike. For illustration, lift height is exaggerated; with this much separation, joint would become unstable.

Figure 2. Separation mechanism, operated by vertical springs and counter-pressure on piston rod side of pneumatic loading cylinder, of HUT-4 hip joint simulator. In this system test with 28 mm AoA, cup inclination angle was  $64^\circ$ , lift height (vertical separation) was 1.0 mm, and lubricant was distilled water. Arrow indicates where spacers are placed for adjustment of vertical separation.

Figure 3. Metasul 50 mm MoM bearing of HUT-4 hip simulator test with swing phase separation. In this photograph separation is 1.0 mm vertically. Bone cement was cast around acetabular component in a hemispherical 68 mm diameter mould that had a recess of 40 mm diameter for formation of flat loading surface (top). As this surface was horizontal in test, position of cup in this case was  $59^\circ$  abduction and  $30^\circ$  anteversion, and so effective inclination was  $64^\circ$ . Femoral neck angle was  $45^\circ$ . Loading direction was vertical.

Figure 4. Load and motion profiles used in HUT-4 tests. FE = flexion-extension, AA = abduction-adduction. Cycle time = 0.94 s.

Figure 5. BHR femoral heads (46 and 54 mm), stems of which have been cut, and tailor-made stainless steel fixation parts with 1:10 taper before cementing. Note polyacetal guide sleeve that prevents stainless steel/CoCr contact and spacer pins. Average thickness of bone cement mantle was 3 mm.

Figure 6. CoCr MoM femoral heads, and head holder made from stainless steel for HUT-4 hip simulator. Left, 54 mm BHR head into which fixation piece was cemented. Right, 52 mm Metasul head. Front, double-taper fixation spigot and 12/14-18/20 head adapter, both made from CoCr. Taper interfaces were sealed from serum by silicone sealant in tests.

Figure 7. Wear test ongoing in HUT-4 hip simulator with three MoM bearings, each of which was immersed in 500 ml of diluted serum contained by acrylic lubricant chamber. Note outer cradle for flexion-extension and inner cradle for abduction-adduction.

Figure 8. Loading beam of HUT-4 hip simulator stiffened for increased loading generated by 80 mm diameter pneumatic double-piston cylinder arrangement capable of 6 kN per test station with 0.7 MPa pressure.

Figure 9. Alumina-on-alumina bearing of 28 mm nominal diameter deliberately roughened with emery paper for friction tests with BRM simulator.

Figure 10. Optical micrograph of streak-like damage by grain removal on 28 mm diameter alumina femoral head caused by impact of superior edge of alumina cup at heel strike in HUT-4 hip simulator test with macroscopic separation of bearing surfaces in swing phase. Vertical separation 1.0 mm, cup inclination 64°, peak load 2 kN, test duration 1 million cycles.

Figure 11. Optical micrograph of streak-like damage by plastic deformation on 50 mm diameter CoCr femoral head caused by impact of superior edge of CoCr cup at heel strike in HUT-4 hip simulator test with macroscopic separation of bearing surfaces in swing phase. Vertical separation 1.0 mm, cup inclination 64°, peak load 2 kN, test duration 1.1 million cycles.

Figure 12. Optical micrographs of scratches with criss-cross pattern attributable to multidirectional sliding, produced on femoral head of 52 mm Metasul MoM in 4 kN test with HUT-4 hip simulator.

Figure 13. Left, BHR cup (54 mm inner dia.) that was in optimal position of 48° inclination in HUT-4 wear test and so entire contact was at a distance from edge of cup, and fluid film lubrication was effective. Inner border of light-coloured protein residue indicates size of contact. Right, BHR cup (46 mm inner dia.) that was in steep 64° inclination and so fluid film lubrication failed and contact was bordered by edge of cup, metal-metal contact took place and wear rate was an order of magnitude higher than in optimal position. In both cases, peak load was 2 kN.

Figure 14. Metal-on-metal bearing of 50 mm nominal diameter damaged by water lubricated friction test in BRM simulator, showing abrasion and galling.

Figure 15. Alumina-on-alumina bearing of 28 mm nominal diameter stained by titanium particles that were deliberately put between bearing surfaces. In HUT-4 trials, squeak occurred in water, but not in serum.



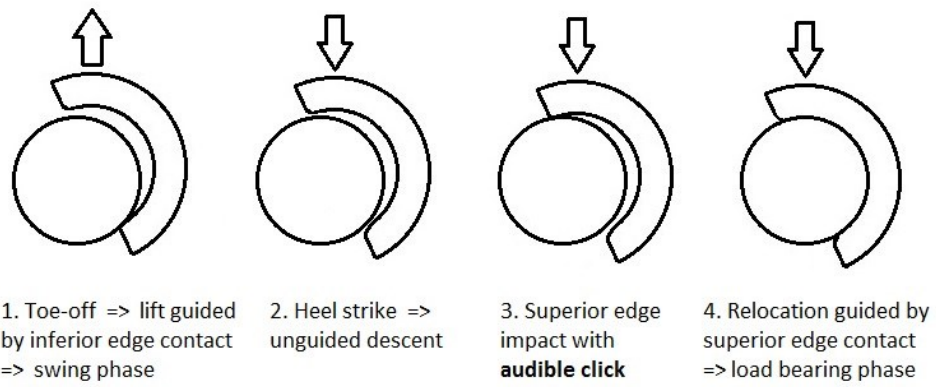


Figure 1.



Figure 2.

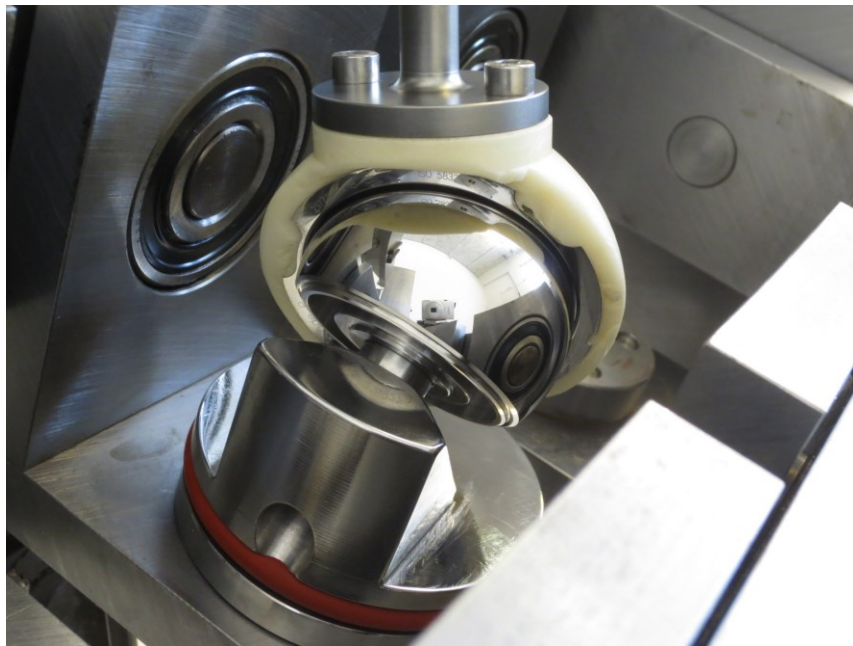


Figure 3.

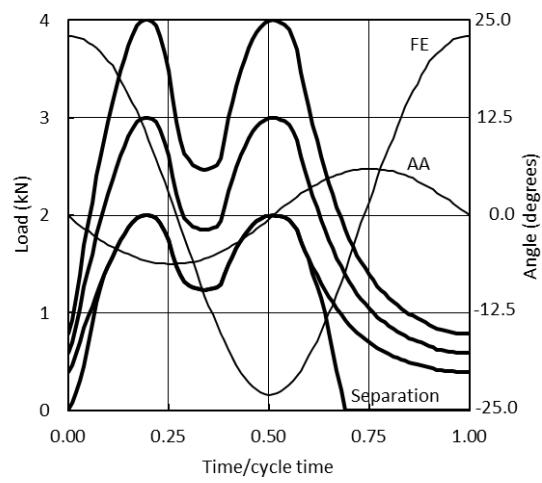


Figure 4.



Figure 5.



Figure 6.



Figure 7.

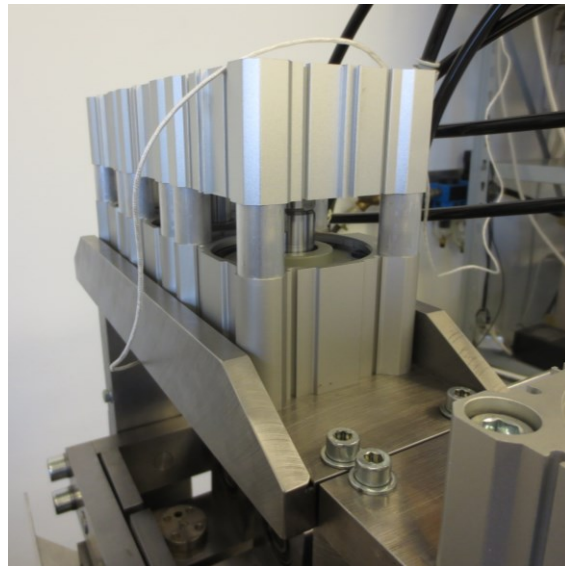


Figure 8.



Figure 9.



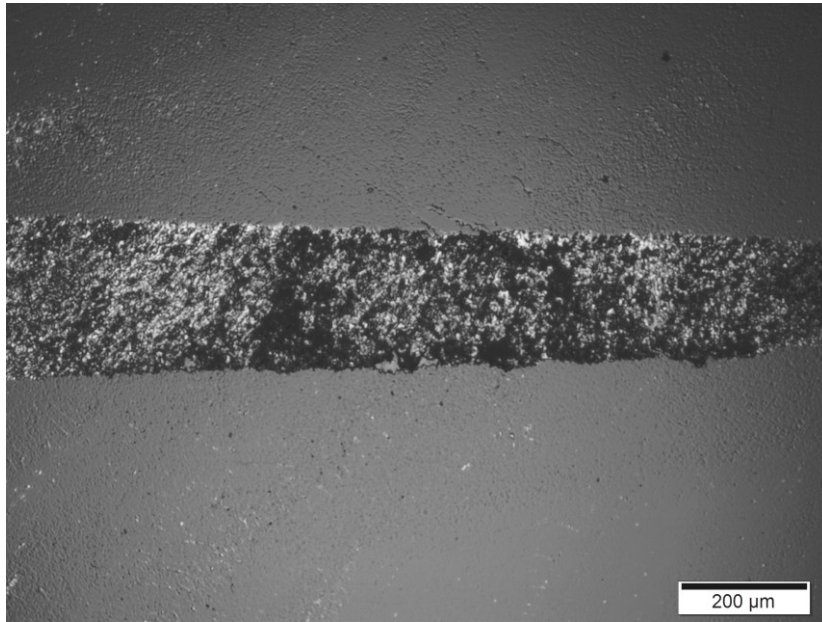


Figure 10.

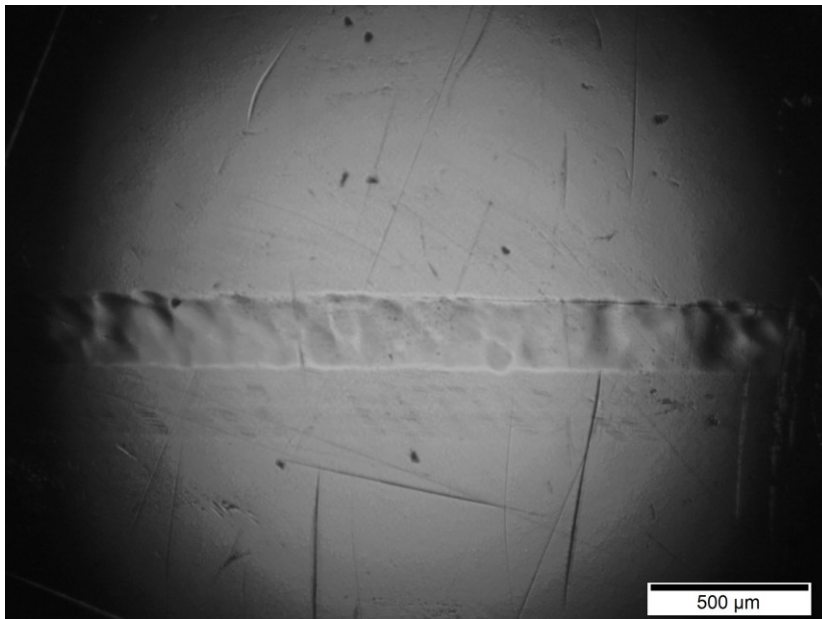


Figure 11.

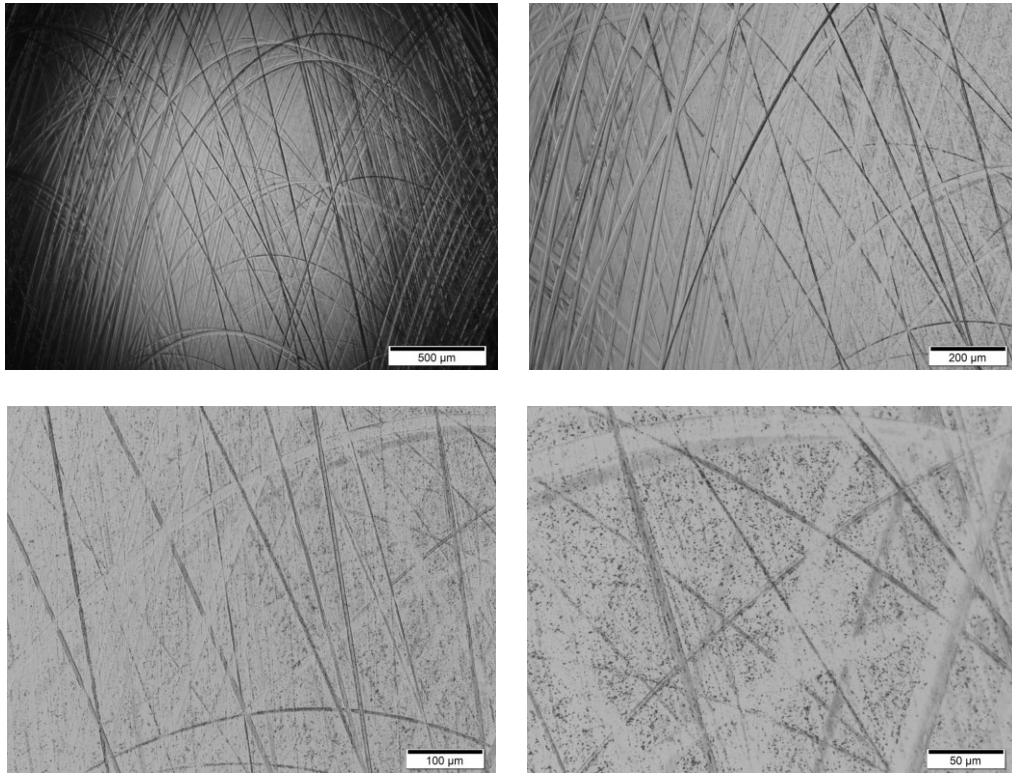


Figure 12.

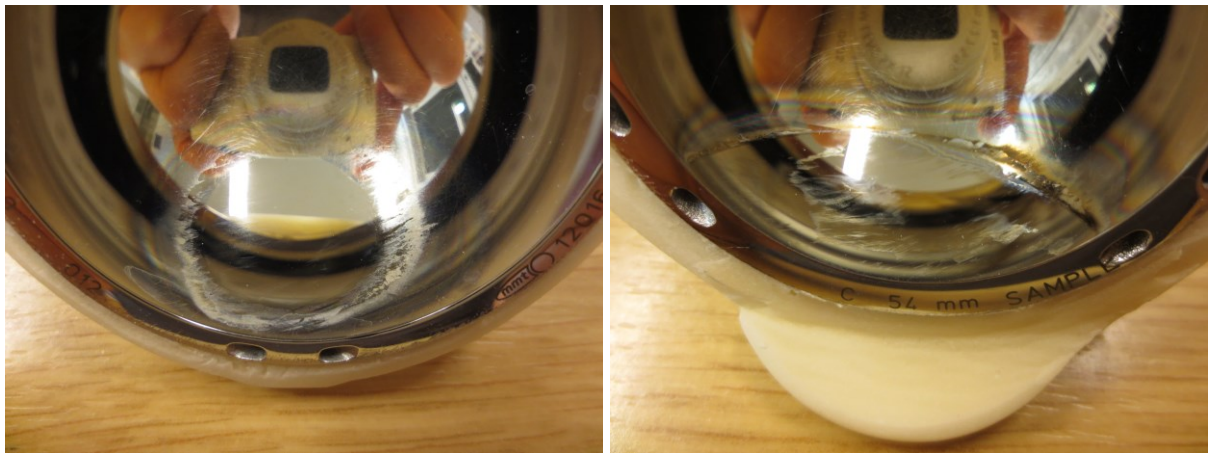


Figure 13.



Figure 14.



Figure 15.